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Muscle strength, task performance and low back load in nurses

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Poor muscle strength, relative to the physical demands of specific jobs, is considered a risk factor for low back pain. To gain an understanding of the underlying mechanisms, this study questioned whether muscle strength was related to task performance and low back load in nursing tasks. Trunk extension, elbow flexion and knee extension strength were therefore measured in 17 nurses. The independent effects of muscle strength on task duration, jerkiness of effort and L5-S1 torque were investigated as the nurses performed several patient handling tasks. Despite a large variation in muscle strength within the subject population, no effect of strength on task duration, jerkiness or L5-S1 torques was observed. In conclusion, poor muscle strength was found not to be related to increased low back load. If 'weaker' nurses were to be at a higher risk, it would be due to a reduced capability to withstand the mechanical load, rather than to an increased mechanical load.

1. Introduction

The nursing profession is associated with a high incidence of low back problems (LBP) (Pheasant and Stubbs 1992). The strenuous physical load associated with patient handling activities generally is considered a main causative factor. This load originates largely from the magnitude of the weight handled, which exceed by far the maximal allowable weights proposed for manual materials handling (Mital *et al*. 1993). Body postures constrained by the working environment and opposing efforts of the patient further augment the physical load. It is likely that in activities such as turning patients in bed or transferring patients from a bed to a chair, the work load is close to the nurse's maximum physical capabilities. As daily exposure to this load becomes more frequent, the risk for LBP increases (Jensen 1990).

It could be argued that nurses are at a higher risk when their muscular strength is lower. Back muscles working closer to their maximum capacity are likely to be more susceptible to injury. Also, weak muscles may result in high peak loads on the spine because of more jerky movement patterns and patterns of force generation. Both muscular damage (Roy *et al*. 1989) and damage to spinal structures (Bogduk and Twomey 1989) have been considered primary causes in LBP development.

The relationship between muscle strength and LBP incidence has been frequently studied. Cross-sectional studies indicate that LBP patients produce lower maximal

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trunk extension torques as compared with healthy persons (e.g. Pope *et al.* 1985). This seems to be mainly due to a reduced muscle capacity caused by inactivity during the LBP period or to a torque production inhibited by the fear of pain. Poor muscle strength as a risk factor for LBP can be only established in a prospective study, wherein the muscle strength of workers is measured and future LBP cases are analysed. Various prospective studies have not shown any or only a poor influence of absolute muscle strength on LBP incidence (Biering-Sörensen 1984, Battié *et al.* 1989). However, in various studies in which the subject's strength relative to physical demands of their jobs was taken into consideration, significant effects were found (Chaffin and Park 1973, Chaffin *et al.* 1978, Cady *et al.* 1979). These studies indicate that high muscle strength can be a protective factor for those who are performing tasks involving high demands in relation to their capacity. As such this finding supports the belief in strength tests as a valuable tool for selecting workers for strenuous jobs.

Two mechanisms may explain how muscle strength affects the risk for LBP: (1) poor muscle strength is related to inefficient task performance and a relatively high mechanical load on low back level; (2) poor muscle strength is related to a decreased capacity of low back structures to withstand the mechanical load. This study examined whether muscle strength affects task performance and, thereby, low back load. Back extension strength, elbow flexion strength and knee extension strength were determined in 17 nurses. These subjects also performed three patient handling tasks. The jerkiness of the effort to perform the tasks and the task duration were determined. Net joint torques at the lumbo-sacral (L5-S1) joint were estimated. Muscle strength effects on these variables were studied in an analysis of co-variance. As body mass was presumably related to strength and it directly affects low back load (simply due to the upper body weight 'resting' on low back level), body mass was treated as a co-variate in the analysis. Thus, muscle strength effects were investigated independently from any potentially obscuring body mass effect. The hypothesis was that nurses with stronger muscles are able to handle patients quicker and with less jerkiness, thereby minimizing the mechanical load on the lumbar spine (lower peak and time-integrated torques), which implies a lower risk for low back injury.

2. Method

Part of the methodology concerning the nursing tasks and the determination of the mechanical load on the low back has been described previously (de Looze *et al.* 1994).

2.1. *Subjects and patient handling tasks*

Nine female and eight male nurses (body height 1.70 ± 0.07 m; total body mass 69.0 ± 8.6 kg) participated in this study. All were free of LBP at the time of the experiments. The patient to be handled was a male (weight 78 kg, height 1.86 m, age 25 years). He was instructed to neither co-operate nor work against the nurse's effort.

The nurses performed three patient-handling tasks: (1) turning the lying patient over from his back to his left side; (2) pulling/lifting the patient from sitting on the edge of the bed so that he is standing on his feet; and (3) lowering the patient from standing position to a sitting position on the edge of his bed. Tasks were performed at a normal steady pace and they comprised movements largely limited to the sagittal plane. Each task was performed at two different bed heights: a fixed height of 0.715 m (i.e. from ground level to the upper side of the mattress), which is a standard bed height in the university hospital in Brussels (AZVUB) and at a bed height preferred by the subjects. The sequence of the tasks was balanced over the subjects.

2.2. *Measurements and biomechanical model*

To estimate the mechanical load on low back level during task performance, a two-dimensional linked segment model was applied (de Looze *et al.* 1992). This model represented five body segments: feet, lower and upper legs, and pelvis and trunk, which were inter-connected at the ankle, knee, hip and L5-S1 joint. Net joint torques were calculated by inverse dynamic analysis. The model required the following input variables: ground reaction forces, kinematic data and anthropometric data about the subject's body segments.

While subjects performed the tasks a force platform (Kistler) recorded the vertical and fore-aft components of the ground reaction force. Analogue force signals were low-pass filtered (30 Hz, 4th order, 24 DB/oct) and sampled (60 Hz).

To analyse the body movements, light reflective markers were placed on the subject's right side at the fifth metatarsophalangeal joint, the distal part of lateral malleolus, the lateral femoral epicondyle, the uppermost margin of greater trochanter, the rotation axis between the fifth lumbar and first sacral vertebra (L5-S1) from a lateral view, the spinous process of the first thoracic (T1) vertebra and the acromion. Marker positions were recorded (60 frames s^{-1}) by a motion analysis system (VICON, Oxford Metrics, Oxford, UK). The marker's coordinates in the sagittal plane were low-pass filtered (effective cut-off frequency of 5 Hz, zero phase lag, 2nd order Butterworth). From the filtered positions the dynamic linked segment model was constructed.

Prior to the experiments, body height, body mass and body segment lengths were measured. These measurements plus data from tables (Winter 1979), were used to estimate the segmental masses, the moments of inertia and the relative positions of the centres of gravity.

The beginning and end points of each trial were determined from the ground reaction force signal. Peak values for the ground reaction force were determined, as was the smoothness of the signal. The smoothness of this signal can be viewed as an index for the overall jerkiness of the nurse's effort to perform the task, as both the acceleration of body segments and the forces generated in patient handling find expression in the ground reaction force. The smoothness was computed by taking the mean of the squared, differentiated force-signal. Similarly, the mean squared differentiated acceleration of the human body has been used to express the jerkiness of motion by Scholz (1993). Finally, the peak and time-integrated (over the total duration of each task) values of the L5-S1 torques were determined.

2.3. *Muscle strength*

Muscle strength tests of the main trunk extensors were performed on a dynamometer (Iso-Station B200, Iso-Technologies, Hillsborough, USA). Subjects were standing and strapped into this apparatus, such that body movements other than trunk flexion and extension were excluded. Prior to testing, the full range of motion from trunk flexion to extension was determined. During the tests, the maximal isometric torque was determined in a standing position in two trials and then averaged. Next, the subject generated a maximal trunk extensor torque by pushing against the dynamometer's backward rotating arm (resistance 50% of the maximal isometric torque) across the total range of motion. The arm's axis was placed at the level of the assumed helical axis localization for the L5-S1 segment. The torque exerted on the axis of the arm was digitized and stored on disk. Each subject performed three trials. The maximal torque value and the average torque over the entire movement were determined and then averaged over all trials.

An iso-kinetic dynamometer (Kincom, Chattecx Corp., Chattanooga, USA) was used to determine the elbow flexion and knee extension strength. Again subjects were strapped into the apparatus; this time maximal torques against a rotating arm could be generated by the elbow flexors or the knee extensors over a range of motion of 90°: for knee extension from 90° flexion to full extension and for elbow flexion from full extension to 90° flexion. This arm rotated at a constant angular velocity of 1.04 rad s^{-1}. Tests were performed for the left and right extremity and under concentric and eccentric conditions; five trials per condition were performed. The maximal and average torque generated during the entire movement were determined for each trial and the values were averaged (over trials).

2.4. *Data analysis*

The correlation between the right and left knee extension and elbow flexion strength was calculated and the results from the left and right extremity were then averaged. Next, correlations between the average and maximal values and between the concentric and eccentric obtained values for elbow flexion and knee extension torques were computed.

Subjects were ranked from low to high for the various torques, i.e. for knee extension and elbow flexion, the maximal concentric, maximal eccentric, average concentric and average eccentric torque and for trunk extension the maximal and average concentric and the maximal isometric torque. On the basis of these rankings, the subjects were finally ranked for the overall elbow, knee and trunk strength and divided for these three parameters into a low, a middle and a high strength group.

Multiple analysis of co-variance with repeated measures and Tukey-HSD contrast tests were used to study the significance ($p = 0.05$) of effects of trunk, elbow and knee strength on the task duration, the ground reaction force (peak and smoothness) and the lumbo-sacral torque (peak and time-integral). To study these effects independently from the variation in body mass, this variable was treated as a co-variate. The type of task and the bed height were treated as within-subject factors; trunk, elbow and knee strength as between-subject factors.

3. Results

3.1. *Muscle strength*

The concentric and eccentric elbow flexion and knee extension torques were averaged over the left and right side after the significance of left versus right correlations had been ascertained (table 1). Figure 1 shows the mean and maximal values of the elbow flexion torque and knee extension torque as measured under concentric and eccentric conditions. Among subjects a large variation was observed for the various strength parameters. For instance, the highest individual value for the maximal concentric elbow flexion torque was 3.4 times higher as compared with the lowest. The eccentric torques generated were always higher than the concentric torques and the correlations between the concentric and eccentric torque values were

Table 1. Correlations between the left and right extremity for the various parameters of elbow flexion and knee extension strength.

	Elbow flexion strength	Knee extension strength
Average, concentric	0.92	0.90
Average, eccenteric	0.86	0.89
Maximal, concentric	0.82	0.93
Maximal, eccenteric	0.80	0.87

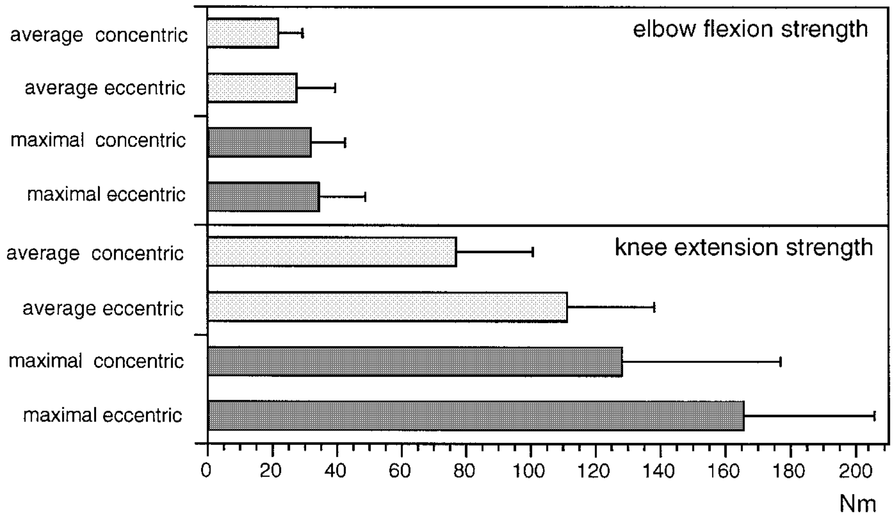

Figure 1. Means and SDs of the maximal and averaged torques in elbow flexion and knee extension. Data from the left and right extremities were averaged.

all statistically significant. Average torque values correlated well with maximal values (table 2).

Figure 2 shows results obtained from the trunk strength dynamometer. The maximal concentric values ranged from 50.9 to 140.0 Nm, the mean concentric values from 38.0 to 101.3 Nm, while the maximal isometric values ranged from 88.0 to 187.7 Nm. The correlation between average and maximal values of the concentric torque was 0.98. The isometric torque however was not significantly correlated to the concentric torque values. Because of this result, it was decided to consider in further analysis the separate effects of isometric and concentric trunk strength, as well as the effect of the overall trunk strength measure.

After ranking subjects for overall trunk extension, elbow flexion and knee extension strength, the correlations among these strength indices was computed. These were 0.71 for elbow versus knee strength, 0.86 for trunk versus elbow strength and 0.80 for trunk versus knee strength.

3.2. *Task performance and low back load*

The task duration was on average 3.8 s (± 1.0) for turning, 2.3 s (± 0.5) for lifting and 2.3 s (± 0.7) for lowering the patient.

Table 2. Correlations between the concenteric and eccentric torques and between the average and maximal torques.

Concentric – eccentric correlation		Average – maximal correlation	
Average elbow flexion	0.68	Concentric elbow flexion	0.95
Maximal elbow flexion	0.60	Eccentric elbow flexion	0.98
Average knee extension	0.88	Concentric knee extension	0.98
Maximal knee extension	0.89	Eccentric knee extension	0.97

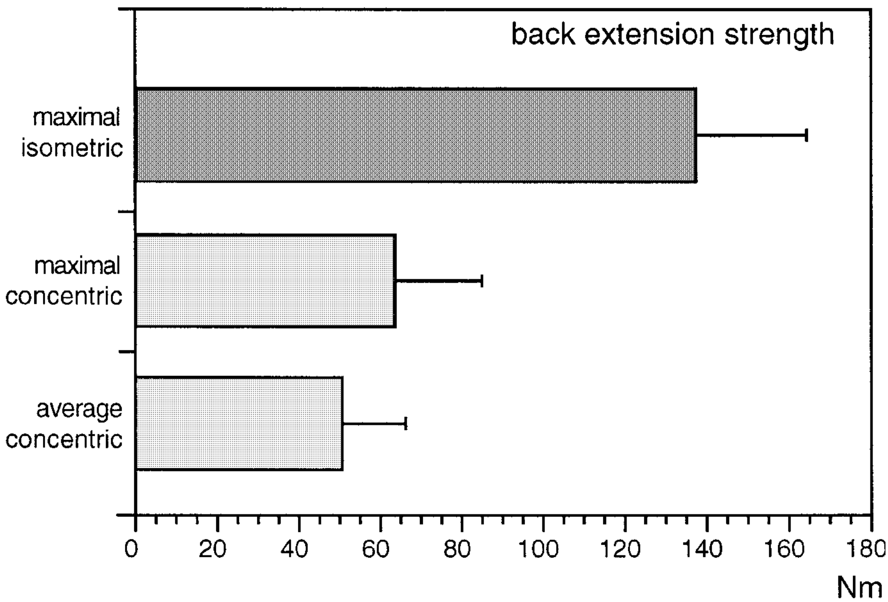

Figure 2. Means and SDs of the maximal isometric torque and the maximal and averaged torques in dynamic back extension.

The ground reaction force is viewed as the combined effect of all the effort to perform a task. The vertical component of this force was found to reach peak values that were on average 21% ($\pm$12), 40% ($\pm$11) and 29% ($\pm$10) above the subject's body mass for the turning, lifting and lowering tasks, respectively. The peak fore-aft ground reaction force was on average 147 N ($\pm$54) in turning, 24 N ($\pm$21) in lifting and 4 N ($\pm$55) in lowering. The jerkiness, deduced from the ground reaction force signal, was significantly higher in the turning task as compared with the lifting and lowering task. No differences between the standard and adjusted bed height conditions were observed with respect to the peak ground reaction forces or with respect to the effort's jerkiness.

Figure 3 shows the results for the different parameters of low back load. The values for peak L5-S1 torques were not significantly different for the various tasks; these were 197 Nm ($\pm$57) in turning, 212 Nm ($\pm$56) in lifting and 197 Nm ($\pm$54) in lowering respectively. With regard to the time-integrals of the L5-S1 torque, the values in the turning task were higher than in the other tasks, because of a longer task duration. More interesting, the effect of bed height on the time-integrated torque was found to be significant: the individually chosen height adjustments resulted in a decrease of the time-integrated torques.

3.3. *Effects of muscle strength on task performance and low back load*

Task performance was not found to be affected by the strength level of the individual nurse. The analysis of co-variance did not reveal any significant effect of elbow, knee or trunk strength on task duration, peak ground reaction force or jerkiness of the

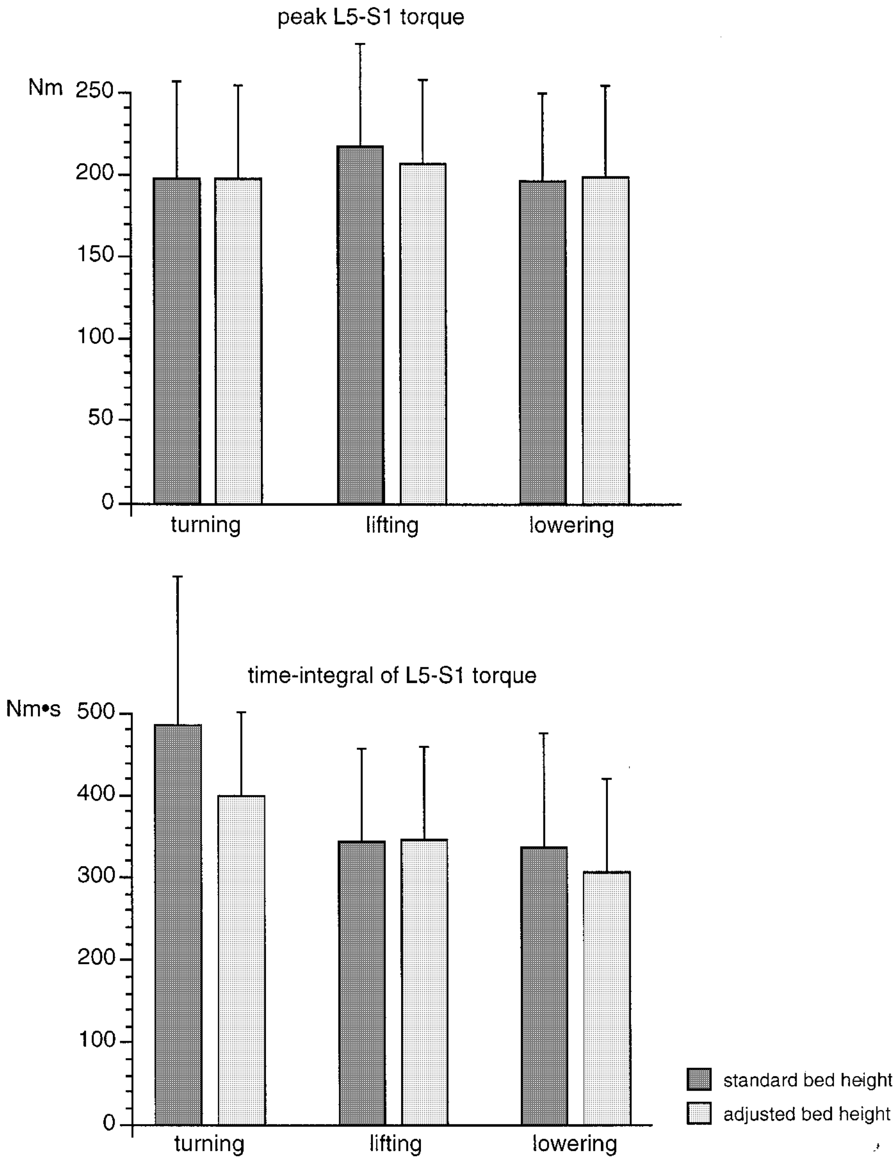

Figure 3. Means and SDs of the peak values and time-integrals of the net joint torque at the lumbo-sacral spinal motion segment.

effort. Furthermore, no muscle strength effect was found on the peak or time-integrated values of the L5-S1 torque. The same result was found when the male and female nurses were treated separately (instead of together) in the statistical analysis. The absence of the hypothesized relationships will be discussed below.

4. Discussion

4.1. *Effects of muscle strength*

The frequently postulated relationship between poor muscle strength and LBP prevalence can be explained by the notion of a reduced muscular capacity to generate force as a symptom of the disease or the notion of poor muscle strength as a LBP risk factor. The latter finds support in the literature when strength is considered in relation to the job demands (Chaffin and Park 1973, Chaffin *et al.* 1978, Cady *et al.* 1979). The rationale behind poor muscle strength as an LBP risk factor was addressed in the present study. It was hypothesized that weaker nurses would be more at risk due to a less efficient task performance and a higher load at low back level. However, no evidence was found for such mechanism. Despite a large variation in muscle strength within the subject population, no effect of strength was found on task performance or low back load. Before conclusions are drawn, the following points are of interest.

It should be stressed that body mass was treated as a co-variate, because body mass was related to strength and because the effect of body mass on low back load is obvious: heavier subjects show higher L5-S1 torques due to more upper body weight. This does not mean that heavier ones are at higher risk as the strength of relevant structures is also positively related with body mass (Hansson and Bengt 1980). In the current study, the independent effect of muscle strength, not obscured by body mass effects, was investigated.

To measure muscle strength specifically, iso-dynamic equipment was used. With regard to trunk strength, however, maximal dynamic torques were found to be well below maximal isometric torques, while maximal isometric torques were well below the torques in nursing. Lower dynamic compared with isometric strength is in line with the previous reports (Burdorf *et al.* 1995). More striking are the higher torques in nursing as compared with the torques in maximal testing. The jerkiness at the onset of the nursing tasks yielding high accelerations of body segments, which are clearly absent in the strength test, may explain this difference. In addition, some systematic error may occur between the directly measured torques in testing and the torques in nursing estimated by biomechanical modelling.

As a parameter of low back load, the lumbo-sacral torque was estimated. This reflects the minimal trunk extension force required and thereby the minimal load on internal body structures (e.g. spinal compression). The distribution of the actual extension force across low back structures may differ among individuals of varying strength, which in turn affect the actual load on internal structures. Co-activation, which is likely to occur in the tasks studied, may further enhance the internal load to a varying extent across individuals with different strength. As long as these features have not been demonstrated nor denied, the existence of any strength effect on the internal low back load cannot be denied with certainty from the present study.

Assuming that the low back load in nursing is not affected by muscle strength, weaker persons might still be more at risk due to a lower capacity to withstand the load. Reduced muscle strength implies a higher intensity of muscle actions to produce the same L5-S1 torque. This increases the rate of homeostatic disturbances within the muscle fibre, which is assumed to play a role in the development of muscle damage and muscular LBP. There is also evidence that poor muscle strength is related to a reduced capacity of passive structures. For instance, Bevier *et al*. (1989) found that isometric back strength was significantly correlated with the bone mineral density (BMD) of spinal vertebrae (which determines the strength of spinal motion segments (Hansson *et al*. 1987). In another study (Snow-Harter *et al*. 1990) the dynamic strength of various muscle groups (back, elbow flexors, leg extensors) and BMD at various sites (femur, spine and mid-radius) was measured and the general conclusion was that dynamic muscle strength could be identified as an independent predictor of BMD.

4.2. *Effects of task and bed height on low back load*

Apart from muscle strength effects, the effect of the task type and bed height was studied. Peak L5-S1 torque did not differ in turning, lifting and lowering of the patient. There was a slight (not significant) tendency for lower values in the lowering as compared with the lifting task. This result agrees with observations in manual handling, showing only minor differences in peak load (4 – 7%) between the lifting and the lowering of materials (de Looze *et al*. 1993).

The tasks were performed on a standard and an individually preferred bed height. Thus it was possible to study whether nurses are capable of reducing the mechanical load on their back by adjusting bed height. The adjustment led to a reduction in the total amount of back load over time without a concomitant rise in peak loads in the turning and lowering tasks, a result which was reported and discussed previously (de Looze *et al*. 1994). Another finding was that the adjustment of bed height chosen by the nurses did not correlate with their individual body height. It was hypothesized that muscle strength could be a determining factor in the chosen bed height. This hypothesis however was not confirmed in this study.

4.3. *Conclusions*

In this study on nursing tasks no relationship was found between various parameters of strength on one hand and parameters of task performance and mechanical low back load on the other. If persons with poor muscle strength would be at a higher risk for LBP as suggested in the literature, it is because of a reduced capability to withstand the mechanical load rather than an increased mechanical load involved in the performance of tasks. As such, strengthening programmes or strength-based worker selection still might help in reducing the LBP prevalency in nurses, although the results found here do not directly favour these strategies. In our opinion, a primary strategy should be the support of the use of mechanical aids like patient hoists, sliding and roller boards or gait and ambulation belts. It has been clearly demonstrated that these aids reduce the low back load in nurses considerably (Garg *et al*. 1991, Garg and Owen 1992).

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References

Battié, M., Bigos, S.J., Fisher, L.D., Hansson, T.H., Jones, M.E. and Wortley, M.D. 1989, Isometric lifting strength as a predictor of industrial back pain reports, *Spine*, **14,** 851 – 856.

Bevier, W.C., Wiswell, R.A., Pyka, G., Kozak, K.C., Newhall, K.M. and Marcus, R. 1989, Relationship of body composition, muscle strength and aerobic capacity to bone mineral density in older men and women, *Journal of Bone Mineral Research*, **4,** 421 – 432.

Biering-Sörensen, F. 1984, Physical measurements as risk indicators for low back trouble over a one-year period, *Spine*, **9,** 106 – 119.

Bogduk, N. and Twomey, L.T. 1989, *Clinical Anatomy of the Lumbar Spine* (Melbourne: Churchill-Livingstone).

Burdorf, A., van Riel, M., van Wingerden, J.P., van Wingerden, S. and Snijders C. 1995, Isodynamic evaluation of trunk muscles and low back pain among workers in a steel factory, *Ergonomics*, **38,** 2107 – 2118.

Cady, L.D., Bisschoff, D.P., O'Connel, E.R., Thomas, P.C. and Allan, J.H. 1979, Strength and fitness and subsequent back injuries in fire-fighters. *Journal of Occupational Medicine*, **21,** 269 – 272.

Chaffin, D.B. and Park, K.S. 1973, A longitudinal study of low back pain as associated with occupational weight lifting factors. *American Industrial Hygiene Association Journal*, **34,** 513 – 525.

Chaffin, D.B., Herrin, G.D. and Keyserling, W.M. 1978, Pre-employment strength testing — an updated position, *Journal of Occupational Medicine*, **20,** 403 – 408.

Garg, A. and Owen, B. 1992, Reducing back stress to nursing personnel: an ergonomic intervention in a nursing home, *Ergonomics*, **35,** 1353 – 1375.

Garg, A., Owen, B., Beller, D. and Banaag, J. 1991, A biomechanical and ergonomic evaluation of patient transferring tasks: wheelchair to shower chair and shower chair too wheelchair, *Ergonomics*, **34,** 407 – 419.

Hansson, T. and Bengt, R. 1980, The influence of age, height and weight on the bone mineral content of lumbar vertebrae, *Spine*, **5,** 545 – 551.

Hansson, T.H., Keller, T.S. and Spengler, D.M. 1987, Mechanical behavior of the human lumbar spine. Fatigue strength during dynamic compressive loading, *Journal of Orthopaedic Research*, **5,** 479 – 487.

Jensen, R.C. 1990, Back injuries among nursing personnel related to exposure, *Applied Occupational Environmental Hygiene*, **5,** 38 – 45.

Looze, M.P. de, Kingma, I., Bussmann, J.B.J. and Toussaint, H.M. 1992, Validation of a dynamic linked segment model to calculate joint moments in lifting. *Clinical Biomechanics*, **7,** 161 – 169.

Looze, M.P. de, Toussaint, H.M., van Dieën, J.H. and Kemper, H.C.G. 1993, Joint moments and muscle activity in the lower extremities and low back in lifting and lowering tasks, *Journal of Biomechanics*, **26,** 1067 – 1076.

Looze, M.P. de, Zinzen, E., Caboor, D., Heyblom, P., van Bree, E., van Roy, P., Toussaint, H.M. and Clarijs, J.P. 1994, The effect of individually chosen bed height adjustments on the low back stress in nurses, *Scandinavian Journal of Work Environment and Health*, **20,** 427 – 434.

Mital, A., Nicholson, A.S. and Ayoub, M. 1993, *Guidelines: Manual Materials Handling* (London: Taylor & Francis).

Pheasant, S. and Stubbs, D. 1992, Back pain in nurses: epidemiology and risk assessment, *Applied Ergonomics*, **23,** 226 – 232.

Pope, M.H., Bevins, T., Wilder, D.G. and Frymoyer, D.W. 1985, The relationship between anthropometric, postural, muscular and mobility characteristics of males, *Spine*, **10,** 644 – 648.

Roy, S.H., de Luca, C.J. and Casavant, A. 1989, Lumbar muscle fatigue and chronic lower back pain, *Spine*, **4,** 992 – 1001.

Scholz, J.P. 1993, Organizational principles for the coordination of lifting, *Human Movement Science*, **12,** 537 – 576.

Snow-Harter, C., Bouxsein, M., Lewis, B., Charette, S., Weinstein, P. and Marcus, R. 1990, Muscle strength as a predictor of bone mineral density in young women. *Journal of Bone Mineral Research*, **5**, 589 – 595.

Winter, D.A. 1979, *Biomechanics of Human Movement* (New York: John Wiley).